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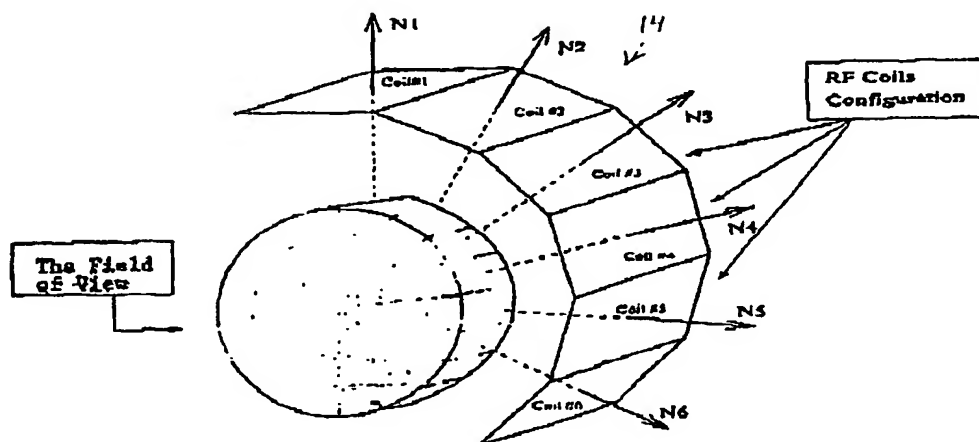
Schematic representation in 3-D of the Field of View and the RF pickup coil array positioning in this method. The coil orientations are described by vectors orthogonal to the coils.

(57) Abstract: Magnetic resonance data is acquired in parallel by an array of separate RF coils (2, 4, 6) arranged circumferentially around the imaging subject. The apparatus and method operate on the basis of determining an estimate of the sensitivity profile of each coil in the array.

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METHOD AND APPARATUS FOR PARALLEL DATA ACQUISITION FROM A MRI COIL ARRAY

BACKGROUND

5   Field of Invention

The present invention generally relates to apparatus and methods for nuclear magnetic resonance imaging (NMRI), also known as magnetic resonance imaging (MRI). More specifically, the invention relates to apparatus and methods for decreasing

10   magnetic resonance (MR) data acquisition times, wherein magnetic resonance data is acquired in parallel using an array of receiver coils at least partially surrounding the object of interest, and the desired MRI image is then reconstructed in parallel.

15   Summary of the Prior Art

In dynamic MRI applications, such as functional imaging, interventional imaging and cardiac imaging, there is has long been a need in the art for methods and apparatus that provide high quality (e.g., high-resolution and signal-to-noise ratio)

20   images. Conventional MRI imaging apparatus and methods, however, operate at speeds that are an order of magnitude slower than those which are currently deemed to be desirable. Some of these conventional methods are described in the Background section of U.S. Patent No. 5,365,172 to Hrovat et

25   al. for "Methods and Apparatus for MRI", the disclosure of which is incorporated herein by reference.

In an attempt to attain faster operating speeds, several so-called "parallel" encoding apparatus combinations and/or methods have been developed. These apparatus combinations

30   and/or methods rely on the use of multiple receiver coils for the acquisition of magnetic resonance data and high-speed data processors for the reconstruction of the field of view with significantly smaller data sets.

Among the parallel imaging techniques described in the literature, the work of Kwiat et al. ("A Decoupled Coil Detector Array for Fast Image Acquisition in Magnetic Resonance Imaging", *Medical Physics*, 18:251, 1991, the

5   disclosure of which is hereby incorporated herein by reference) is significant. This work involved the investigation of methods for solving the inverse source problem on magnetic resonance signals received in multiple RF receiver coils. The technique proposed required the use of a

10   number of RF coils equal to the number of pixels in the desired image. It also required that the sensitivity of the coils used be increased by an order of magnitude. Since these requirements are quite impractical in conventional magnetic resonance imaging (wherein the usual number of pixels in the

15   image is on the order of 256x256), this technique has never been used successfully in a biological imaging experiment.

Other so-called "parallel" imaging techniques that use one dimensional sensitivity profiles of RF coils to encode space in a MRI context also have been proposed. For example,

20   Ra, et al. ("Fast Imaging Using Sub-encoding Data Sets From Multiple Detectors", *Magn. Reson. Med.*, 30:142, 1993, the disclosure of which is hereby incorporated by reference herein), describes a method that uses sets of equally spaced k-space lines from multiple receiver coils in a line, and

25   combines them with the one dimensional sensitivity profile information to remove the aliasing that occurs due to undersampling. A four-fold decrease in the image acquisition time of a water phantom was postulated to be possible by using an array of four coils.

30   Nevertheless, no biological images were shown in this article. It is believed that this may be indicative of a possible lack of robustness of the alaising removal algorithm in practical situations.

A method called SMASH proposed by Sodickson et al.
("Simultaneous Acquisition Of Spatial Harmonics (SMASH): Fast
Imaging With Radio Frequency Coil Arrays", *Magn. Reson. Med.*,
38:591-603, 1997, the disclosure of which is hereby
5   incorporated herein by reference) has been found to be
practical, and to yielded good results. SMASH enhances
imaging speed by using multiple RF receiver coils. More
specifically, it uses linear combinations of the 1D
sensitivity profiles of receiver coils (weighted so as to form
10   sinusoidal harmonics) of a one dimensional array to generate
all k-space lines from a small subset of collected magnetic
resonance data.

The SMASH method, however, is somewhat limited. It has
an inherent inflexibility in the choice of the imaging plane
15   to be viewed. Also, it has a demonstrable limitation in depth
penetration. Further, there is a practical, physical limit on
the number of coils that can be placed along one direction in
a magnetic resonance imaging apparatus - particularly if the
coils are to be de-coupled from one another.
20

SUMMARY OF THE INVENTION

The present invention provides an apparatus and a method
that significantly decrease both magnetic resonance data
acquisition time, and image reconstruction time, in magnetic
25   resonance imaging.

Also, the present invention provides an apparatus and a
method wherein sets of magnetic resonance information are
acquired simultaneously, in parallel, with one another, and
the elements of those sets preferably are subsequently
30   processed in parallel with one another to reconstruct images.

These, and other, features and advantages of the present
invention constitute a generalization of, and an improvement
upon, the SMASH apparatus and method. More specifically, the
present invention contemplates the placement of an array of RF

coils, comprising substantially any number of RF receiver
coils, at least partially around an object of interest located
in the imaging volume of a magnetic imaging device. The
present invention also contemplates the provision of a so-

5   called "parallel" imaging capability wherein the output image
may be taken in any plane transverse to the imaging volume of
the apparatus.

In the present invention, parallel encoding in MRI is
achieved by using the sensitivity profiles of an array of RF

10   receiver coils at least partially surrounding the object of
interest. Given this fact, the equation describing the MR
signal seen by the ith coil of a coil array surrounding the
imaging volume of a magnetic resonance imaging device can be
written as:

15

$$S_i(t) = \int \int \rho(x,y).W_i(x,y).e^{j\gamma(G_x xt + G_y y\tau)} dx.dy$$

Where W_i (x,y) represents the 2D sensitivity profile of the ith
coil of the array, and $\rho(x, y)$ is an image slice in a selected

20   (x, y) plane. Then, taking the Fourier transform of that
signal with respect to x yields:

$$F_i(x) = FT[S_i(t)] = \int \rho(x,y).W_i(x,y).e^{j\gamma G_y \tau}.dy$$

25   The latter equation represents a projection of the phase
modulated image $\rho(x,y)$ onto the x-axis. Further, this signal
can be represented in discrete form by the following matrix
product:

$$F_i(x) = \Sigma_y [W_i(x,y).e^{j\gamma G_y \tau}].[\rho(x,y)]$$

30

If the number of receiver coils used simultaneously is N, and
the 2D sensitivity profile $W_i(x,y)$ of each one of them is

4

known, it is possible to reconstruct the image from only one N^{th} of the total number of k-space lines that would normally be required. Accordingly, the apparatus and method of the present invention use phase modulated projections of the

5   received magnetic resonance data onto the frequency encoded (x-) axis, weighted by the 2D sensitivity profiles of the coils in the array, in order to reconstruct $\rho(x,y)$ column by column (i.e., orthogonal to the x-axis).

The 2D sensitivity profiles are calculated first by

10 acquiring a baseline image using the RF body coil of the conventional magnetic resonance imaging apparatus, whereby the sensitivity profile may be considered to be constant ($W_B(x,y)$ = 1) and a slice of the image in the (x, y) plane may be written as: $\rho(x,y)$. An image is then acquired with each coil in the

15 array. In the ith coil, this image can be written as:

$$W_i(x,y)\ \rho(x,y)$$

$W_i(x,y)$ is then computed by forming a point-by-point ratio between magnetic resonance data from the ith coil and that from the body coil.

20   It, therefore, will be seen that a preferred embodiment of the invention is a method for generating a magnetic resonance image of an object of interest composed of a plurality of adjacent image lines. The method generally includes the following steps. First, a magnetic resonance

25 imaging device is provided. This device typically includes a magnet system providing a background magnetic field in an imaging volume, central processor means, memory means, a RF coil surrounding the imaging volume, and a plurality of radio frequency receiver coils defining a multi-dimensional array

30 thereof disposed about the imaging volume. A two-dimensional sensitivity profile for each receiver coil in the multi-dimensional array is then computed and recorded to the memory means. Thereafter, a plurality of magnetic resonance signals

of the object of interest located within the imaging volume is acquired from each receiver coil and recorded to the memory means. An image of the object of interest in a desired plane extending transversely through the imaging volume of the

5   magnetic resonance imaging device then is reconstructed line-by line by the central processor means. This reconstruction combines the inverse of the matrix of the sensitivity profiles of each receiver coil and the matrix of the recorded MR data signals together to provide an image that may be displayed or

10   printed.

Preferably, the method of the invention involves a calibration step wherein a homogeneous water phantom is provided to obtain the sensitivity profile of each receiver in the array. The homogeneous water phantom is located in the

15   image volume of the magnetic resonance imaging device. Magnetic resonance data from the homogeneous water phantom in each plane of interest extending transversely through the imaging volume of the magnetic resonance imaging device then are acquired by a RF body coil and by each of the RF image

20   receiver coils. This magnetic resonance data is stored in the memory of the magnetic resonance imaging device. Thereafter, the central processor means calculates, and stores in the memory means, the respective point-by-point ratios of the complex data representing the image from the water phantom

25   provided by said receiver coils to the complex data representing the image from the water phantom provided by the RF body coil.

In a preferred embodiment of the invention, an object of interest is located in the imaging volume of the magnetic

30   resonance imaging device. Magnetic resonance data from the plane of interest is acquired from each of the RF receiver coils and stored in memory. Then, the respective point-by-point ratios of the magnetic resonance data from the plane of interest provided by the receiver coils to the magnetic

resonance data from the water phantom are calculated so as to define a scaling factor for each point in the plane of interest.   This scaling factor may then be utilized to provide a load-weighted, point-by-point sensitivity profile for each

5   RF receiver coil in each plane of interest parallel to that for which the scaling factor was originally determined.

In still another embodiment of the invention, means are provided for determining and storing in memory the point-by-point time difference of signal reception by each of the

10   receiver coils.   This information may be combined with the sensitivity information in the inverse of the matrix of the sensitivity profile so as to provide a phase compensated, load-weighted, point-by-point, inverse, sensitivity profile matrix for each said receiver coil.

15   Further, it will be understood that the present invention includes apparatus for carrying out each of the above-described methods.   Accordingly, the present invention also includes an apparatus for generating a magnetic resonance image of an object of interest - the image being composed of a

20   plurality of adjacent image lines.   More specifically, the apparatus preferably constitutes a magnetic resonance imaging device including an imaging volume, preprogrammed central processor means, memory means, a body coil surrounding said imaging volume, a plurality of radio frequency receiver coils

25   defining a multi-dimensional array thereof disposed about said imaging volume, and display means.   Each receiver coil in the multi-dimensional array has an associated two-dimensional sensitivity profile.   Preferably, the preprogrammed central processing means is adapted to calculate each of the

30   sensitivity profiles and to transfer the same to the memory means.

The receiver coils are adapted to acquire simultaneously a plurality of magnetic resonance signals from an object of interest located within the imaging volume, and to transfer

the same to the memory means. In addition, the central
processor means is adapted to reconstruct, and to display on
the display means, line-by-line, a two-dimensional image taken
in a selected plane extending transversely through the object
5   of interest.

This is accomplished by multiplying the inverse of the
matrix of sensitivity profiles of the receiver coils and the
matrix of data signals acquired by the receiver coils, and
displaying the resultant product on the display means.

10 In another embodiment, the magnetic resonance imaging
device is adapted to acquire, and to transfer to the memory
means, magnetic resonance data from a homogeneous phantom in
each plane of interest extending through the imaging volume
using said body coil. Similarly, the magnetic resonance
15 imaging device also is adapted to acquire, and to transfer to
the memory means, magnetic resonance data from a homogeneous
water phantom in each plane of interest from each of said
receiver coils. Also, the central processing means is adapted
to calculate, and transfer to the memory means, the respective
20 point-by-point ratios of the magnetic resonance data from the
water phantom provided by the receiver coils to the magnetic
resonance data from the water phantom provided by the body
coil. The component points of each image line may then be
refined in the central processing means by a corresponding one
25 of these ratios prior to its display on the display means.

In yet another embodiment, the magnetic imaging device is
adapted to acquire, and to transfer to the memory means,
magnetic resonance data from the object of interest by each of
the receiver coils. In this case, the central processing
30 means is adapted to calculate, and to transfer to the memory
means, the respective point-by-point ratios of the magnetic
resonance data from the object of interest provided by the
receiver coils to the magnetic resonance data from the water
phantom provided by the receiver coils. This ratio defines

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for each point a scaling factor. The central processing means also is adapted to refine the component points of each image line by use of a corresponding one of the scaling factors prior to its display.

5   Finally, in still another embodiment of the invention, phase detection means are provided for determining and storing in the memory means the point-by-point time difference of signal reception by each of the receiver coils. In this alternative, the central processing means is adapted to
10   multiply the point-by-point time difference of signal reception by the point-by-point sensitivity profile for each said surface coil prior to the line-by-line image display on the display means.

15   BRIEF DESCRIPTION OF THE DRAWINGS

These and other features, objects and advantages of the present invention, will be more completely understood by those skilled in the art from the following detailed description of the preferred embodiments of the invention read in conjunction
20   with the appended drawings. In the appended drawings, like reference numeral are used to refer to like elements throughout, and:

Fig. 1 is an illustrative depiction of a magnetic resonance imaging device in accordance with the present
25   invention;

Fig. 1A is a schematic illustration of the disposition of the RF receiver coil array of the present invention disposed at least partially surrounding the imaging volume of the magnetic imaging device shown in Fig. 1;

30   Fig. 2 is a flow chart illustratively depicting the steps of the present method and their interrelationship with one another;

Fig. 3 is a schematic representation of the MRI acquisition and image reconstruction scheme of the present invention;

Fig. 4 is a schematic illustration of an exemplary one-
5   dimensional version of the acquisition and image reconstruction scheme of the present invention;

Figs. 4A-4C are schematic illustrations of the sensitivity profiles denoted by $W_1(i, j)$, $W_2(i, j)$ and $W_3(i,j)$, respectively;

10   Fig. 5 is a composite image depicting simulation images received with a coil having the same sensitivity profile placed at 16 different positions about a brain;

Fig. 6 is a composite image showing the results of a reconstruction simulation using different sets of k-space
15   lines;

Figs. 7A-7B illustratively depict reconstructions of a (256x256) image of a GE quality phantom using the body coil and receiver coils with a 16 fold speed enhancement; and

Figs. 8A-8B illustratively depict reconstructions of a
20   brain image using the same techniques as the corresponding depictions of Figs. 7a and 7b, respectively.

DETAILED DESCRIPTION OF PREFERRED EMBODIMENT

25   As generally noted above, the present invention is implemented in a MRI setting. A parallel acquisition apparatus and method using 2D sensitivity maps of receiver coils disposed in an array at various arbitrary locations about an imaging volume in a magnetic resonance imaging
30   apparatus is provided in order to encode the image space.

However, the invention allows for a more general positioning of the RF receiver coils, and can be used to create images derived from any plane in the imaging volume.

Further, the present invention has demonstrated an order
of magnitude gain in imaging speed that varies, depending upon
the number of coils used in the receiver array. In addition,
an easy way of computing the 2D sensitivity profile of a RF

5   receiver coil is provided using a sensitivity map model that
may be acquired from a readily available homogeneous water
phantom.

Hence, the apparatus and method of this invention are
very flexible in that they can be used to speed up any pulse

10   sequence. Indeed, it is anticipated that this novel apparatus
and method will be found to be particularly well suited for
cardiac imaging applications where fast spin-echo sequences
are needed.

Referring now to the drawings, and particularly to Figs.

15   1 and 1A, it will be seen that the present invention provides
a magnetic resonance imaging system that is basically of the
type well known in the art. The system includes a plurality
of gradient coils that produce spatially encoded gradients
imposed upon a background magnetic field B_0 within the volume

20   in which the object to be examined is placed. In addition, an
array of RF receiver coils is disposed in circumferentially
spaced relation to one another about the imaging volume.

More specifically, the magnetic resonance imaging system,
generally indicated at 2, may include a Helmholtz coil pair 4

25   used to generate a large, static, substantially homogeneous
magnetic field in the imaging space or volume 6 in a direction
parallel to the z-axis or the line 8. An object or subject
(not shown) may be placed in the imaging volume within the
cylinder 10 for examination using NMR phenomena. The subject

30   is placed on the z-axis or the line 8, and is located within
the coil 12. The coil 12 is representative of devices used to
generate RF fields in the subject placed in the system for
examination.

These RF fields, when in the presence of a static magnetic field B_0 cause the occurrence of magnetic resonance in the nuclei of certain elements, such as hydrogen, in the specimen to be examined.

5   The direction of the static magnetic field (B_0) produced by the coil pair 4 is indicated in Fig. 1 by an arrow. Currents are made to flow in the RF coil 12. The direction of the currents reverse each half-cycle of the alternating RF current in coil 12. This produces a transverse magnetic field

10   of low magnitude compared to B_0. The magnitude of the flux density resulting from the static magnetic-field intensity B_0 may be typically on the order of Tesla.

The static magnetic field B_0 is constant while the subject is in the system for analysis or examination. The RF

15   transverse magnetic field is applied for a time sufficient only to allow the protons in the hydrogen atoms (or the nuclei of other atoms exhibiting the magnetic resonance phenomenon) to be affected such that precession of the net magnetization of the subject occurs. The precision of the net magnetic

20   field associated with the nuclei in the subject occurs at the Larmor frequency, which is directly proportional to the magnitude of the magnetic field at the location of the nuclei. This can be detected as an NMR signal which provides the information for reconstruction of an image.

25   In MRI systems, it is necessary to provide various gradient coils (not shown) for producing spatially encoding gradients that are imposed upon the static magnetic field within the region in which the subject to be examined is placed also are provided. The gradient-coil apparatus is

30   typically positioned on the outside of a cylindrical surface, such as the surface 10, which may be used as a support structure for the gradient coils. The gradient coils conventionally produce linear magnetic field gradients in any of the three orthogonal directions x, y and z.

Conventionally, the direction of the line 8 is designated as
the z-direction or z-axis and the x- and y-axes of the
coordinate system are orthogonal to the z-axis and to one
another.

5   The typical configuration of a z-gradient coil is
illustrated in U.S. Pat. No. 4,468,622, the disclosure of
which is hereby incorporated by reference. The configuration
of the typical transverse (x or y) gradient coils is
illustrated in U.S. Pat. No. 4,486,711, the disclosure of
10   which also is hereby incorporated by reference.

          The present invention also contemplates that an array of
RF receiver coils 14 will surround (or partially surround) the
imaging volume (see Fig. 1A). These RF receiver coils 14
provide the required system calibration and input information
15   necessary to enhance the speed of parallel magnetic resonance
data acquisition and parallel image reconstruction, without
serious adverse effect upon acceptable signal-to-noise ratios.
More simply, the concept of parallel imaging in the present
invention is based on using multiple receiver coils, with each
20   coil providing independent information about the image.

          Fig. 2 is a flow diagram illustratively showing the basic
sequence of the method of the invention. Thus, it will be
seen that the present method contemplates that an apparatus
such as that described above will be provided. In this
25   apparatus, the CPU is preprogrammed to carry out numerous
operations upon signals received in parallel from various
coils of the receiver array, and to output an image to the
display means. Among these numerous operations are the
following: (1) the computation of the 2D sensitivity profile
30   of each coil of the receiver array; (2) storing the inverse of
the calculated sensitivity profile matrix of each coil of the
receiver array in memory; and (3) reconstructing an image
taken in a selected plane extending transversely through the
imaging volume by calculating and transmitting to the display

means in parallel with one another the various display lines
that make up the entire image.

The theory behind this method has been alluded to above.
Specifically, an MR signal received in a receiving coil having

5   $W_k(x,y)$ as its two-dimensional sensitivity profile, when
neglecting T2 decay, can be written as:

$$s_k(G_y^g, t) = \int \int \rho(x,y) W_k(x,y) e^{i\gamma(G_x xt + G_y^g y\tau)} dx dy$$

10

Where $\rho(x, y)$ denotes the proton density function, G_x
represents the readout gradient amplitude applied in the x
direction, G_y^g represents the phase encoding gradient applied
during the gth acquisition, x and y represent the x and y

15 positions respectively, and τ is the pulse width of the phase
decoding gradient G_x.

In most conventional serial imaging sequences, the phase
decoding gradient is constant along one direction, and the
phase encoding is applied along an orthogonal direction. In

20 addition, only one receiver coil is used to collect all the
data required to reconstruct $\rho(x, y)$. To achieve that, the
phase encoding gradient G_y is varied so as to cover all of k-
space with the desired resolution. For each value of G_y^g of
G_y, an echo is acquired. This makes serial imaging a time-

25 consuming procedure. In this technique, however, sensitivity
profile information from a number of receiver coils is used in
order to minimize the number of acquisitions needed to
estimate and reconstruct $\rho(x, y)$. Taking the Fourier
Transform of equation (1) along the x direction when a phase

30 encoding gradient G_y^g is applied yields:

$$S_k(G_y^g, x) = \int \rho(x,y) W_k(x,y) e^{j\gamma(G_y^g y\tau)} dy,$$

This is the phase-modulated projection of the sensitivity-
weighted image onto the x-axis. A two-dimensional orthogonal
spatial basis function denoted by $\Omega(x, y)$ is used to expand
the quantities in equation (2) for the purpose of discretizing
5   it. A constant spatial sampling dwell of Δ_s is assumed, and
the following change of variables is made:

$$x = m\Delta_s, \quad and \quad y = n\Delta_s.$$

10   Then, performing the expansion in terms of $\Omega(x, y)$ yields:

$$S_k(G_y^g, m\Delta_s) = \int \Sigma_n \rho(m\Delta_s, n\Delta_s).W_k(m\Delta_s, n\Delta_s).e^{i\gamma(G_y^g n\Delta_s \tau)}\Omega(x - m, y - n)dy \quad (4)$$

Removing Δs and rearranging the terms for simplification,
15   results in:

$$S_k(G_y^f, m) = \Sigma_n \rho(m, n).W_k(m, n).e^{i\gamma(G_y^g n\tau)}$$

This expression can be converted to matrix form for each
20   position m along the horizontal direction of the image, as
follows:

$$
\begin{pmatrix}
S_1(G_y^1, m) \\
S_1(G_y^f, m) \\
S_2(G_y^1, m) \\
S_2(G_y^f, m) \\
\vdots \\
S_k(G_y^1, m) \\
S_k(G_y^f, m)
\end{pmatrix}
=
\begin{pmatrix}
W_1(m,1).\phi(G_y^1,1) & \cdots & W_1(m,n).\phi(G_y^1,n) \\
W_1(m,1).\phi(G_y^f,1) & \cdots & W_1(m,n).\phi(G_y^f,n) \\
W_2(m,1).\phi(G_y^1,1) & \cdots & W_2(m,n).\phi(G_y^1,n) \\
W_2(m,1).\phi(G_y^f,1) & \cdots & W_2(m,n).\phi(G_y^f,n) \\
\vdots & & \vdots \\
W_k(m,1).\phi(G_y^1,1) & \cdots & W_k(m,n).\phi(G_y^1,n) \\
W_k(m,1).\phi(G_y^f,1) & \cdots & W_k(m,n).\phi(G_y^f,n)
\end{pmatrix}
\cdot
\begin{pmatrix}
\rho(m,1) \\
\rho(m,2) \\
\rho(m,3) \\
\vdots \\
\vdots \\
\rho(m,n)
\end{pmatrix}
$$

30

Where f is the number of phase encodes used in the experiment,

$$\phi(G_y^g, p) = e^{i\gamma(G_y^g p \tau)}$$ for every position p along the y axis

and every phase encoding gradient value $G^g{}_y$.

Equation (6) can be written in a more compact form as follows:

5

$$[S_k(G_y^g, m)] = [W_k(m, n).e^{i\gamma(G_y^g(n)\tau)}].[\rho(m, n)]$$

Where $[S_k(G^g{}_y, m)]$ represents a matrix of dimensions $(K \times N)$, and
K is the product of the number of coils k in the array by the
number of phase encodes acquired f, which is equal to the

10   number of echoes collected in all the coils combined; M is the
number of time points per echo, which is equal to the number
of pixels resolved in the x direction, and N is the desired
number of pixels in the y direction. For square image
matrices, M must equal N.

15   The matrix $[A_m(k, y)] = [W_k(m, n).ei\gamma(Gy(n)\tau)]$ is three-
dimensional with dimensions $(K \times M \times N)$, and $[p(m, n)]$ represents
the image of size $(M \times N)$. If the matrix index m is fixed to a
value m_j, the size of $[A_{mj}]$ then becomes $(K \times N)$. Hence, in
order to reconstruct the image $[p(m, n)]$, a matrix

20   pseudoinverse of $[A_{mj}(k, y)]$ is computed for every position m_j
along the x-axis. This yields a column by column
reconstruction. The choice of K affects the quality of the
reconstruction. Practically, increasing K would result in an
increase of the rank of the matrix $[A_{mj}(k, y)]$, yielding a

25   pseudoinverse that is better conditioned. Equation 10 is
schematically represented in Fig. 3 for clarity.

       In addition, a simplified version of the above concepts
is depicted in Fig. 4, and described hereinbelow, to further
clarify the present method.

30   Assume the image to be of size 3x3 as shown in Fig. 4,
where each pixel has an intensity value denoted by r(i,j),
where i is the column number and j is the row number. The
goal is to determine and reconstruct the values of r(i,j) for
all the pixels of the image. Further assume that three coils

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1, 2 and 3 are arrayed around the image, and are used for
image acquisition as shown in Fig. 4. These coils have
sensitivity profiles denoted by $W_1(i,j)$, $W_2(i,j)$, $W_3(i,j)$,
respectively. (see Figs.4A-4C)

5   The sensitivity profiles of the coils are determined from
a homogeneous image by placing a water phantom whose pixel
intensity values are equal to $I(i,j)$ into the imaging volume,
and acquiring its image using any chosen pulse sequence with
the body coil whose sensitivity profile $W_b(i,j) = 1$ for all the
10   pixels in the image. The reconstructed image would have pixel
intensity values of $r_b(i,j) = I(i,j)$ for all pixels. Next,
images are acquired using the 3 coils shown in Fig. 4 with
respective sensitivity profiles $[W_1(i,j)]$, $[W_2(i,j)]$, $[W_3(i,j)]$
to get three images of pixel intensity values equal to
15    $[I(i,j).W_1(i,j)]$, $[I(i,j).W_2(i,j)]$, $[I(i,j).W_3(i,j)]$,
respectively. Taking the point-by-point ratio of the images
acquired using the three coils in the array divided by the RF
body coil image yields the sensitivity profiles $W_1(i,j)$,
$W_2(i,j)$, and $W_3(i,j)$, respectively.
20   Thereafter, the homogeneous water phantom is replaced
with the object of interest to be imaged. This object has
pixel intensities $r(i,j)$ as shown in Figure 4. The MR signal
received in coil k can be written as:

$$S_k(t) = \int \int r(x,y).W_k(x,y).e^{j\gamma(G_x xt + G_y y\tau)}dx.dy$$

Where G_x and G_y represent the gradients applied in the x and y
directions respectively. Then taking the fourier transform
30   with respect to x provides:

$$s_k(x) = \int r(x,y).W_k(x,y).e^{j\gamma(G_y y\tau)}dy$$

This is the phase modulated projection of the image r (x,y) weighted by the sensitivity profile $W_k(i,j)$ onto the x-axis. Discretizing, wherein the index i corresponds to a discrete x position and the index j corresponds to a discrete y position,

5   provides:

$$s_k(i) = \Sigma_j r(i,j).W_k(i,j).e^{j\gamma(G_y j\tau)}$$

whereby in coil 1,

10

$$s_1(i) = r(i,1).W_1(i,1).e^{j\gamma(G_y 1\tau)} + r(i,2).W_1(i,2).e^{j\gamma(G_y 2\tau)} + r(i,3).W_1(i,3).e^{j\gamma(G_y 3\tau)}$$

In coil 2,

15   $$s_2(i) = r(i,1).W_2(i,1).e^{j\gamma(G_y 1\tau)} + r(i,2).W_2(i,2).e^{j\gamma(G_y 2\tau)} + r(i,3).W_2(i,3).e^{j\gamma(G_y 3\tau)}$$

and in coil 3,

$$s_3(i) = r(i,1).W_3(i,1).e^{j\gamma(G_y 1\tau)} + r(i,2).W_3(i,2).e^{j\gamma(G_y 2\tau)} + r(i,3).W_3(i,3).e^{j\gamma(G_y 3\tau)}$$

20

These equations represent the modulated projections of the image onto the x axis. They also are three linear equations with three unknowns [r(i,1), r(i,2),r(i,3)] for every position i along the x axis.

25   Recalling that the goal is to solve for the pixel intensity values of the vector [r(i,1), r(i,2),r(i,3)] which is a column of the image at position i along the x axis, it is noted that the following equation may be written for each position i along the x axis:

30

$$\begin{pmatrix} s_1(i) \\ s_2(i) \\ s_3(i) \end{pmatrix} = \begin{pmatrix} W_1(i,1).e^{j\gamma(G_y 1\tau)} & W_1(i,2).e^{j\gamma(G_y 2\tau)} & W_1(i,3).e^{j\gamma(G_y 3\tau)} \\ W_2(i,1).e^{j\gamma(G_y 1\tau)} & W_2(i,2).e^{j\gamma(G_y 2\tau)} & W_2(i,3).e^{j\gamma(G_y 3\tau)} \\ W_3(i,1).e^{j\gamma(G_y 1\tau)} & W_3(i,2).e^{j\gamma(G_y 2\tau)} & W_3(i,3).e^{j\gamma(G_y 3\tau)} \end{pmatrix} . \begin{pmatrix} r(i,1) \\ r(i,2) \\ r(i,3) \end{pmatrix}$$

Subsequently, solving for the vector

$$
\begin{pmatrix} r(i,1) \\ r(i,2) \\ r(i,3) \end{pmatrix}
$$

5   one takes the inverse of the matrix

$$
\begin{pmatrix} W_1(i,1).e^{j\gamma(G_y 1\tau)} & W_1(i,2).e^{j\gamma(G_y 2\tau)} & W_1(i,3).e^{j\gamma(G_y 3\tau)} \\ W_2(i,1).e^{j\gamma(G_y 1\tau)} & W_2(i,2).e^{j\gamma(G_y 2\tau)} & W_2(i,3).e^{j\gamma(G_y 3\tau)} \\ W_3(i,1).e^{j\gamma(G_y 1\tau)} & W_3(i,2).e^{j\gamma(G_y 2\tau)} & W_3(i,3).e^{j\gamma(G_y 3\tau)} \end{pmatrix}
$$

and multiplies it with the vector

10

$$
\begin{pmatrix} s_1(i) \\ s_2(i) \\ s_3(i) \end{pmatrix}
$$

with the result that:

15

$$
\begin{pmatrix} r(i,1) \\ r(i,2) \\ r(i,3) \end{pmatrix} = \begin{pmatrix} W_1(i,1).e^{j\gamma(G_y 1\tau)} & W_1(i,2).e^{j\gamma(C_y 2\tau)} & W_1(i,3).e^{j\gamma(G_y 3\tau)} \\ W_2(i,1).e^{j\gamma(G_y 1\tau)} & W_2(i,2).e^{j\gamma(G_y 2\tau)} & W_2(i,3).e^{j\gamma(G_y 3\tau)} \\ W_3(i,1).e^{j\gamma(G_y 1\tau)} & W_3(i,2).e^{j\gamma(G_y 2\tau)} & W_3(i,3).e^{j\gamma(G_y 3\tau)} \end{pmatrix}^{-1} \begin{pmatrix} s_1(i) \\ s_2(i) \\ s_3(i) \end{pmatrix}
$$

20

Hence, the image may be reconstructed column by column in
parallel.

     As described above, the inventive method is based on
25 using the sensitivity profiles of RF receiver coils to encode
MR data. As suggested in the above simplified example, it is
possible to calculate these profiles by first acquiring a
baseline image of a homogeneous water phantom using a RF coil
with a homogeneous sensitivity profile covering the whole
30 image. This image can be written as $I(x,y)$. Subsequently,
images of the same water phantom are acquired using the
receiver coils. The image acquired using the kth receiver coil
can be represented as $I(x,y).W_k(x,y)$ where $W_k(x,y)$ is the
sensitivity profile of the surface coil.

Then, taking the point-by-point ratio of the two images
yields the sensitivity profile $W_k(x,y)$. It, however, must be
understood that the sensitivity profiles of the receiver coils
also depends upon their loading.

5   In the present invention, the variations incurred by
these profiles amounts to a constant scaling between any two
different loads. Consequently, to find the actual sensitivity
profile of the receiver coils when loaded with an arbitrary
object of interest, sensitivity profile estimation on that

10   object as described above is performed. Then, a comparison
(ratio) of those sensitivity profiles to the sensitivity
profiles computed on the basis of a homogeneous water phantom
is made in order to extract the scaling factor. The
sensitivity profiles calculated from the homogeneous water

15   phantom are then multiplied by the scaling factor so computed,
and the result is used in the encoding scheme. This is done
in order to get the maximum coverage possible of the field of
view by the sensitivity profile. Other methods have been
described in the literature for performing a sensitivity

20   profile calculation. However, the technique described and
illustrated above appears to be particularly well suited for
our reconstruction method. This is particularly true since
the method is not time-consuming, and it can be carried out
essentially in real time.

25   In order to appropriately account for all of the
frequencies included in the image, the choice of the phase
modulations used in the inversion matrix should be determined
by the frequency content of the sensitivity profile. In most
cases, the sensitivity profiles of RF coils have low frequency

30   content. Accordingly, high frequency phase modulations should
be used to well represent the image. This means that
appropriate k-space lines can be chosen by those skilled in
the art for an optimal reconstruction.

Examples

20

Experiments were performed on a 1.5T GE SIGNA clinical MRI
system at the 5.4 hardware-software configuration. The
purpose of these experiments was to test the performance of
the present methodology. In the experiments, image resolution
5   was 256x256 pixels. A 16-echo RARE sequence, TR=2sec, TE=30ms
per shot, were utilized to obtain one dynamic image update
with a single shot.

In order to assess the technique herein described,
simulations were performed.

10   In one case, a homogeneous MR data of a brain first was
acquired using a head coil. Then, two further sets of MR data
were acquired of a homogeneous water phantom - one using the
head coil, and the other using a surface coil. The
sensitivity profile of the receiver coil was then computed by
15   taking the point-by-point ratio of the water phantom data
acquired using the receiver coil over the data acquired using
the head coil, as described above. Thereafter, sensitivity
profiles were computed for RF receiver coils disposed about
the object of interest with an angle of $2\pi/16$ between any
20   adjacent two of them. Next, the computed sensitivity profiles
were multiplied point-by-point with the brain data to get
approximations of sensitivity weighted images from surface
coils placed at different positions. These images are shown
in Fig.5.

25   The sensitivity-weighted images were then Fourier
Transformed in two dimensions in order to get the relevant k-
space data. Subsequently, 16 lines of k-space data were taken
from each matrix, and those lines were used to reconstruct the
head image using our technique. As mentioned, the choice of
30   the k-space lines affects the image resolution. In Fig.6,
four images acquired using different sets of k-space data are
shown. The image on the left shows a reconstruction using the
16 lines of k-space going between k=-7 and k=8; the second
image from the left shows a reconstruction using the 16 lines

of k-space going between k=-39 and k=40 and acquiring every 5th
line . The third image from the left shows a reconstruction
using the lines of k-space going between k=-79 to k=80, and
acquiring every 10th line.

5

Finally, the image on the right shows a reconstruction using
the 16 lines of k-space going between k=-127 to k=128, and
acquiring every 16th line. It can be seen that the third image
from the left represents the best. result of the four shown.

10   Further, a certain deterioration of the image is observed as
more lines are skipped before reconstruction. An optimal
combination of k-space lines can be found to achieve the
desired result.

In another experiment, baseline data of the human head

15   was acquired using the head coil. Then, using a receiver
coil, 16 data sets from the human head taken at different coil
positions were acquired. These data sets then were used in
conjunction with the baseline data to calculate the
sensitivity profiles of the coil at different positions using

20   the point-by-point ratio method described above. These
sensitivity profiles were then used in order to reconstruct
images using only 16 lines of k-space from each coil. The
results of these experiments are shown respectively in Fig.7
and Fig.8.

25   Accordingly, it will be understood that a parallel
imaging and reconstruction method has been designed and
implemented which is intended to substantially increase
imaging speed in MRI. This technique is based on using 2D
sensitivity profile maps derived from an array of receiver

30   coils surrounding an imaging volume in order to encode and
reconstruct MRI data in parallel. Therefore, imaging time may
be reduced in proportion to the number of receiving coils
present in the array. Further, this technique is adaptable to

all imaging sequences, and to all imaging planes, thereby making it very practical in a clinical setting.

Still further, the placement of the RF receiving coils may be arbitrary, around the imaged field of view. This

5 allows for the placement of more RF coils in the receiver array, which in turn enables faster imaging. Nevertheless, since there is a tradeoff between coil size and penetration, an optimal number of coils can be computed to yield the best results.

10 Still further, in most parallel imaging schemes, decoupling of the RF receiver coils in the array is a necessary step before imaging. In the present invention, however, the coupling information between the different coils is incorporated into the reconstruction algorithm. This makes

15 it easier to design and execute arrays that contain a large number of coils.

In the foregoing discussion, the sensitivity profiles of the respective receiver coils have been suggested to be independent of the coil loading. This permits an

20 extrapolation of the results of the sensitivity profiles computed from a water phantom to be used in the encoding of MR data acquired from other objects. In reality however, loading affects the sensitivity profiles of the coil elements. Accordingly, the dynamic method of sensitivity profile

25 adjustment for each different object to be imaged required in the course of the generation of images having practical utility.

It is to be understood that this description of certain presently preferred embodiments of the present invention has

30 been provided by way of illustration, and not limitation. Various alterations, changes, variations, modifications and the like encompassed within the scope of the present invention in its broadest aspects will occur to those skilled in the art in view of the teachings and descriptions contained in this

specification. For example, the array of receiver coils may be used to emit the RF signal to eliminate the need for a RF body coil. Accordingly, the scope, content and limitations of the present invention are to be determined solely from the

5   appended claims.

Claims:

What is claimed is:

1.   A method for generating a magnetic resonance image

5   of an object of interest wherein said image consists of a
plurality of adjacent image lines, said method comprising the
steps of:

a)   providing a magnetic resonance imaging device
comprising a magnet system providing a homogeneous magnetic

10   background field in an imaging volume, a RF body coil
surrounding said imaging volume, and a plurality of radio
frequency receiver coils defining a multi-dimensional array
thereof disposed about said imaging volume;

b)   determining a two-dimensional sensitivity profile

15   for each said receiver coil in said multi-dimensional array;

c)   acquiring simultaneously a plurality of magnetic
resonance signals from said object of interest located within
said imaging volume from each said receiver coil;

d)   reconstructing a two-dimensional image taken in a

20   selected plane extending through said object of
interest using said two-dimensional sensitivity
profile and said plurality of magnetic resonance
signals.

25   2.   The method according to Claim 1, wherein:
step (b) further comprises:

(i)   providing a homogeneous water phantom;

(ii) locating said homogeneous water phantom in said
image volume;

30   (iii)   acquiring magnetic resonance data from
said homogeneous water phantom in each plane of
interest extending transversely through said
imaging volume using said RF body coil;

(iv) using a two-dimensional Fourier Transformation to provide complex image data representing an image of said homogeneous water phantom in each said plane for the body coil;

5   (v) acquiring magnetic resonance data from said homogeneous water phantom in each plane of interest extending transversely through said imaging volume using each of said RF receiver coils;

10   (vi) using a two-dimensional Fourier Transformation to provide complex image data representing an image of said homogeneous water phantom in each said plane for each reciver coil; and

(vii)   determining the respective point-by-point
15   ratios of said complex image data from said water phantom provided by said receiver coils to said magnetic resonance data from said water phantom provided by said body coil, thereby obtaining a sensitivity map for each receiver
20   coil.

3.   The method according to Claim 2, wherein step (b) further comprises:

25   (viii)   locating an object of interest in said imaging volume;

(ix) acquiring magnetic resonance data from said object in each plane of interest extending transversely through said imaging volume using
30   said RF body coil;

(x) using a two-dimensional Fourier Transformation to provide complex image data representing an image of said object in each said plane for the body coil;

(xi)   acquiring magnetic resonance data from said
        object in each plane of interest extending
        transversely through said imaging volume using
        each of said RF receiver coils;

5   (xii)   using a two-dimensional Fourier
        Transformation to provide complex image data
        representing an image of said object in each
        said plane for each receiver coil; and

        (xiii)   calculating in said central processor and
10   storing in said memory the respective point-by-
        point ratios of said complex image data from
        said object of interest provided by said
        receiver coils to said complex image data of
        said object of interest from said body coil to
15   define for each said point a scaling factor.

4.   An apparatus for generating a magnetic resonance image of
an object of interest, wherein said image is composed of a
plurality of adjacent image lines; said apparatus comprising:
     a magnetic resonance imaging device comprising a magnet

5   system providing a homogeneous magnetic background field in an
imaging volume, preprogrammed central processor means, memory
means, a RF body coil surrounding said imaging volume, a
plurality of radio frequency receiver coils defining a multi-
dimensional array thereof disposed about said imaging volume,

10   and display means;
     wherein each said receiver coil in said multi-dimensional
array has an associated two dimensional sensitivity profile;
said preprogrammed central processing means is adapted to
calculate each said sensitivity profile and to transfer same

15   to said memory means;
     wherein said receiver coils are adapted to acquire
simultaneously a plurality of magnetic resonance signals of
said object of interest located within said imaging volume and
to transfer same to said memory means; and

20   wherein said central processor means is adapted to re-
construct, and to display on said display means, line-by-line,
a two dimensional image taken in a selected plane extending
transversely through said object of interest by using said
sensitivity profile of each said coil and corresponding

25   signals from each of said receiver coils.

5.   An apparatus according to Claim 5, wherein said magnetic resonance imaging device further is adapted to acquire and to transfer to said memory means:

5

(i)   magnetic resonance data from a homogeneous water phantom in each plane of interest extending transversely through said imaging volume using said body coil, and

(ii)   magnetic resonance data from said homogeneous water phantom in each said plane of interest

10

from each of said receiver coils;

said central processing means is adapted

(1)   to calculate and transfer to said memory means the respective point-by-point first ratios of complex image data of said water

15

phantom provided by said receiver coils to corresponding complex image data of said water phantom provided by said body coil.
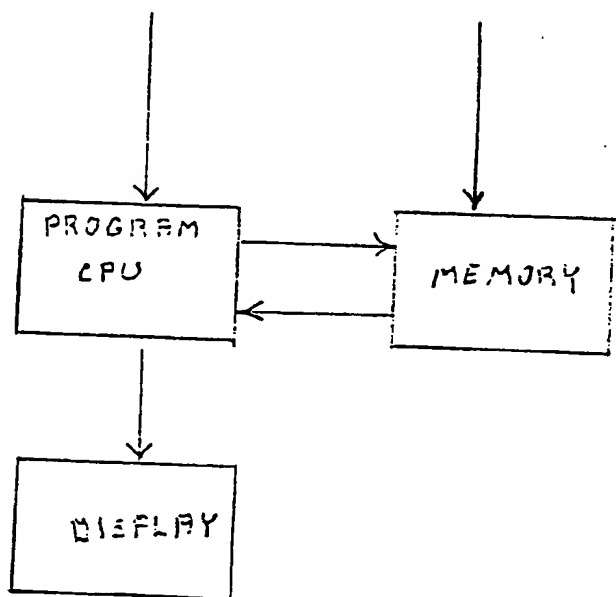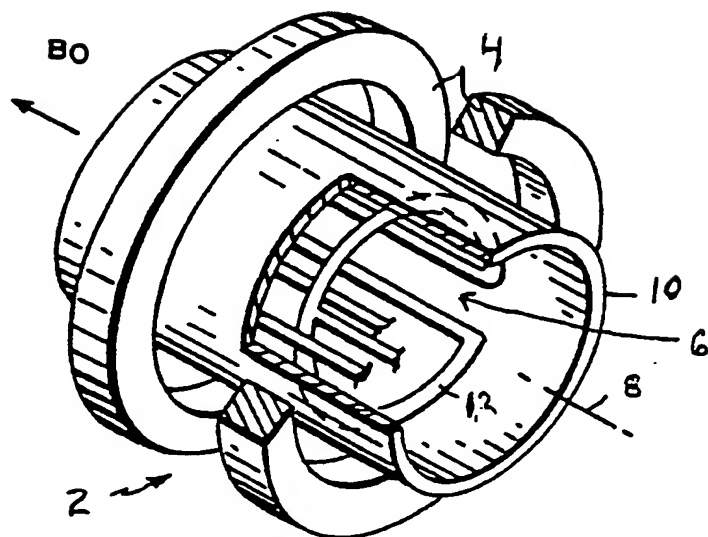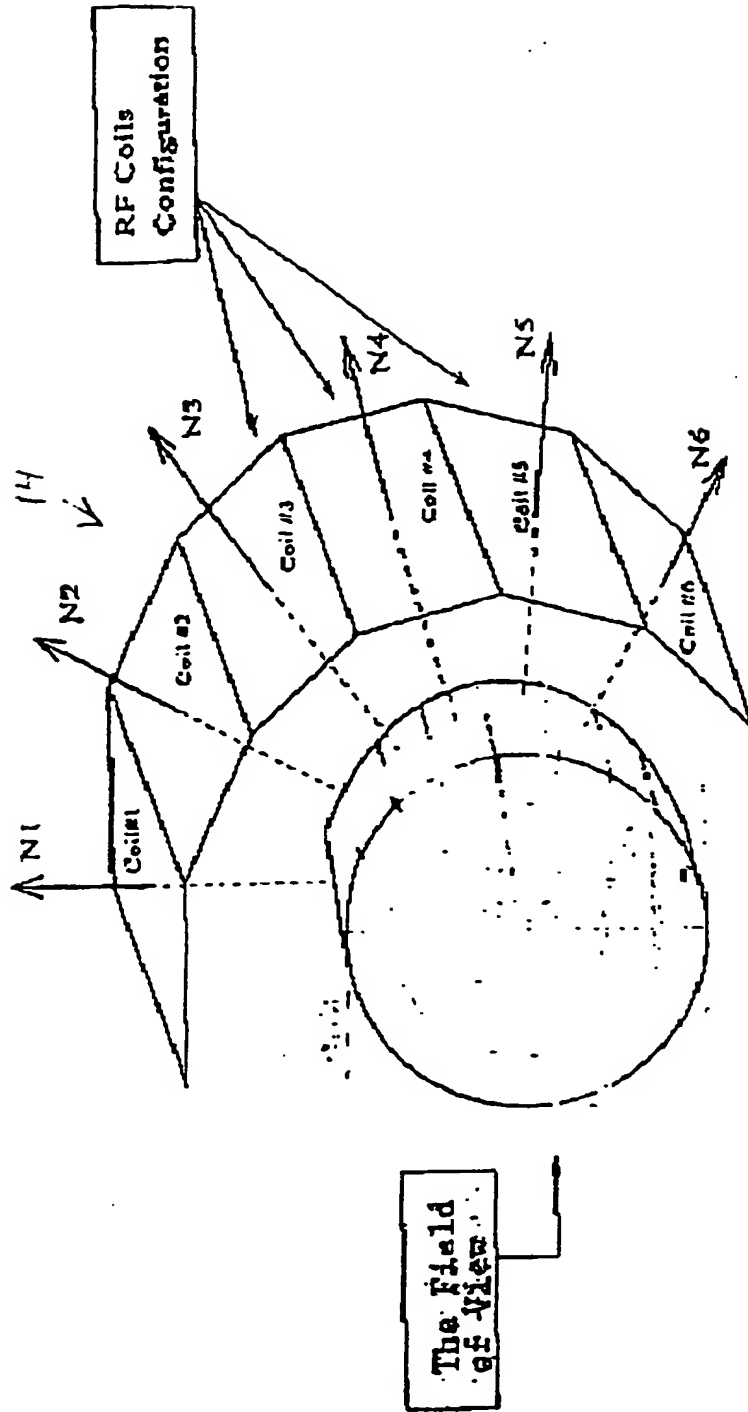
FIG.1

Figure 1A Schematic representation in 3-D of the Field of View and the RF pickup coil array positioning in this method. The coil orientations are described by vectors orthogonal to the coils.

```
┌─────────────────────┐
│  PROVIDE APPARATUS  │
│  AS SHOWN IN FIG. 1 │
└──────────┬──────────┘
           │
           ▼
┌─────────────────────┐                    ┌─ ─ ─ ─ ─ ─ ─ ─ ─ ─ ┐
│ COMPUTE SENSITIVITY │                      REFINE SENSITIVITY
│    2D PROFILE FOR   │ ─ ─ ─ ─ ─ ─ ─ ─ ─▶     PROFILES
│   EACH RECEIVER     │                    │                     │
│      COIL 14        │                    └─ ─ ─ ─ ─ ─ ─ ─ ─ ─ ┘
└──────────┬──────────┘                               │
           │                                          │
           ▼                                          │
┌─────────────────────┐                               │
│   STORE INVERSE     │                               │
│    SENSITIVITY      │ ◀ ─ ─ ─ ─ ─ ─ ─ ─ ─ ─ ─ ─ ─ ─ ┘
│    PROFILES OF      │
│   RECEIVER COILS    │
└──────────┬──────────┘
           │
           ▼
┌─────────────────────┐              ┌─────────────────────┐
│  ACQUIRE MAGNETIC   │              │   STORE ACQUIRED    │
│  RESONANCE DATA     │              │     MAGNETIC        │
│  FROM OBJECT IN     │─────────────▶│  RESONANCE DATA     │
│  RECEIVER COILS IN  │              │    IN MEMORY        │
│     PARALLEL        │              │                     │
└──────────┬──────────┘              └──────────┬──────────┘
           │                                    │
           ▼                                    │
┌─────────────────────┐              ┌──────────┴──────────┐
│ RECONSTRUCT IMAGE   │              │      DISPLAY        │
│ LINES IN PARALLEL   │─────────────▶│      IMAGE         │
└─────────────────────┘              └─────────────────────┘
```
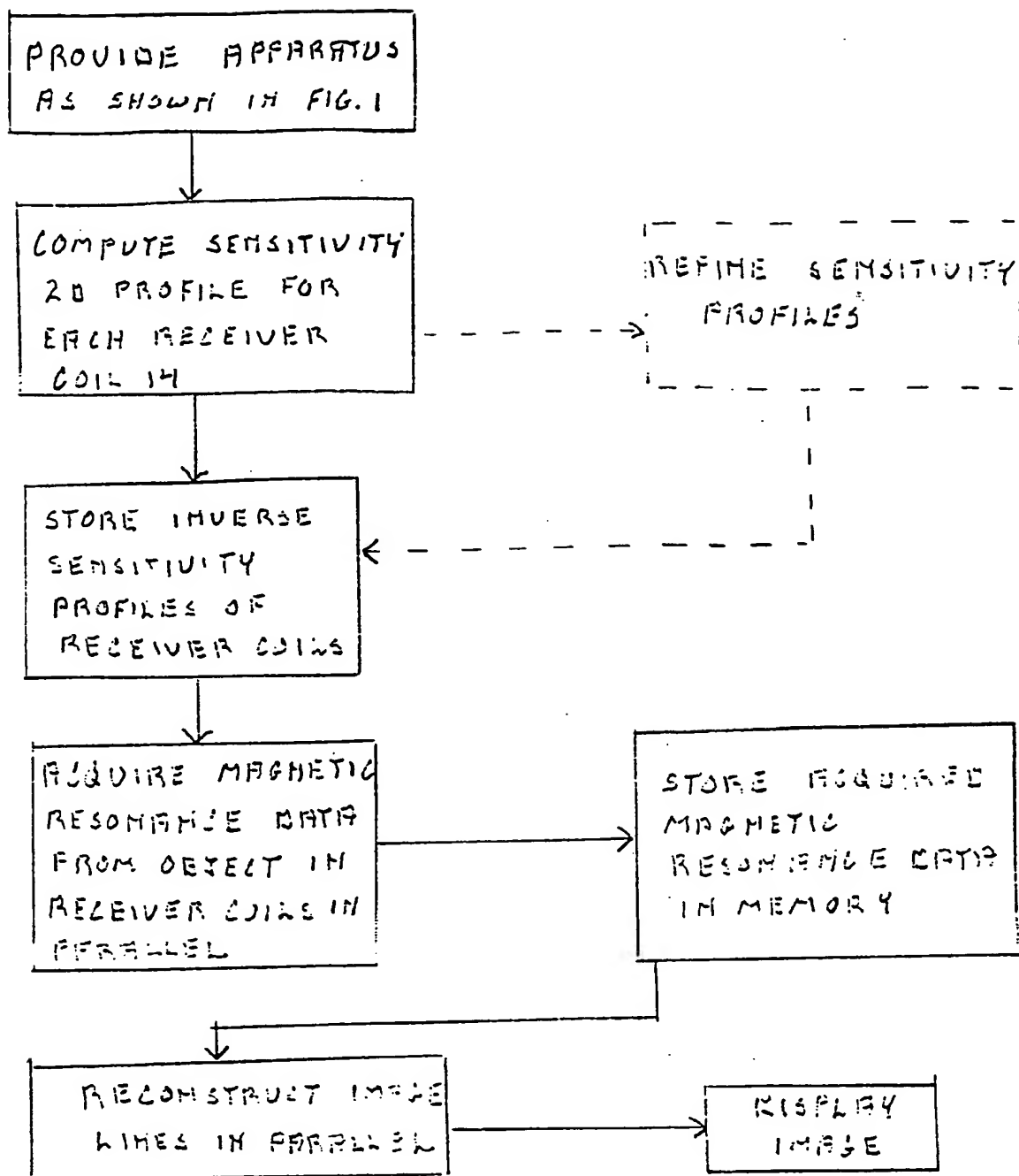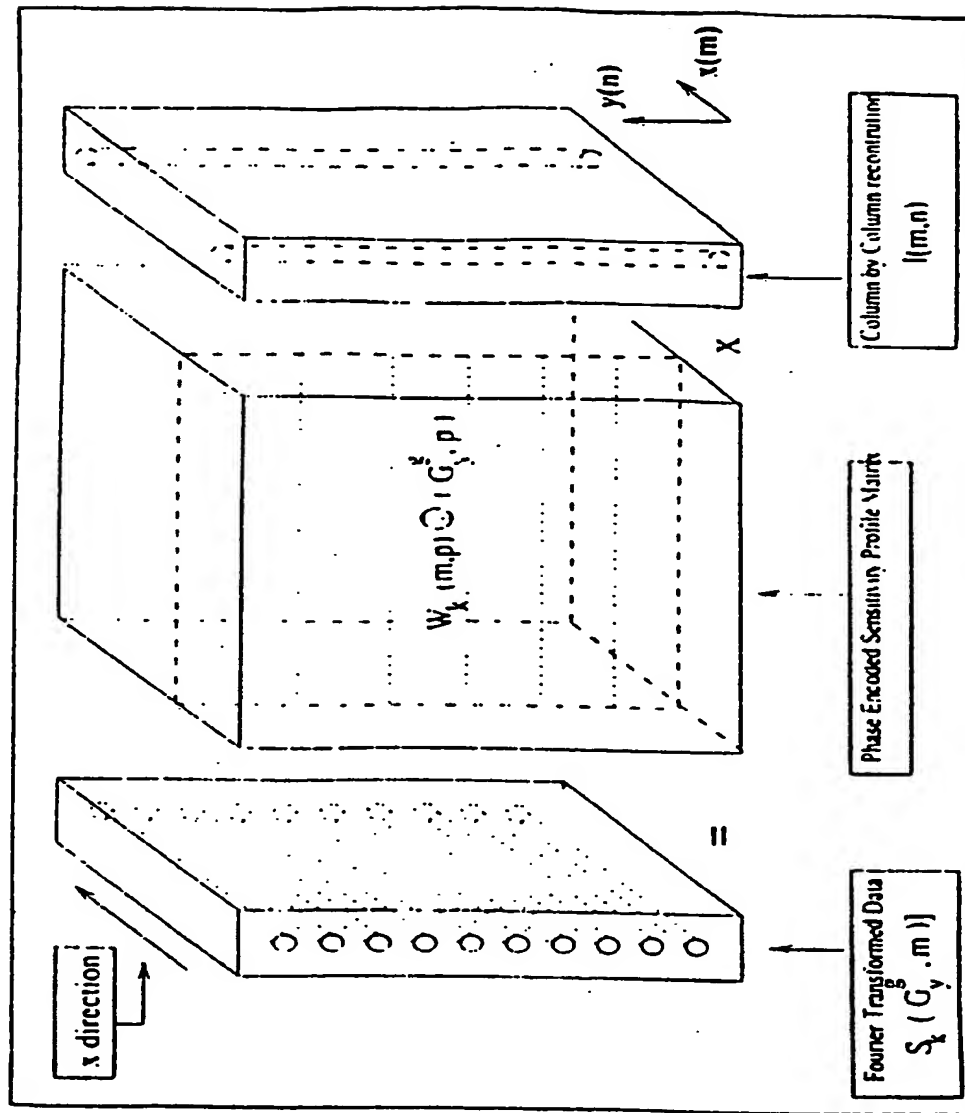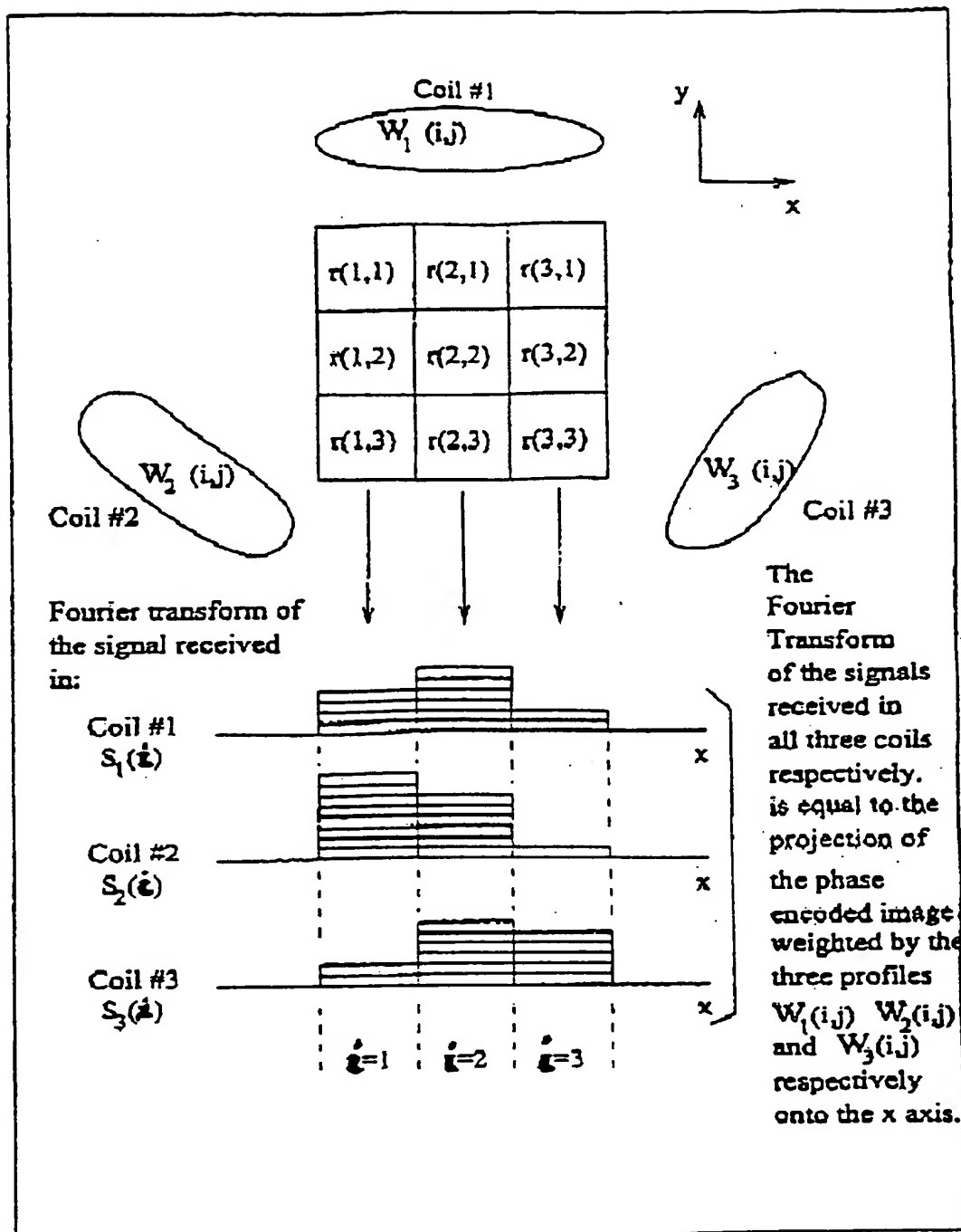
Fig. 2

Figure 3 Schematic representation of the parallel reconstruction scheme. The matrix to the left represents the 1DFT of the k-space data chosen for reconstruction, the matrix in the center is the three dimensional sensitivity matrix referred to by A in the text, and the matrix on the right represents the image to be reconstructed.

Figure 4

Coil #1

$W_1(1,1)$	$W_1(1,2)$	$W_1(1,3)$
$W_1(2,1)$	$W_1(2,2)$	$W_1(2,3)$
$W_1(3,1)$	$W_1(3,2)$	$W_1(3,3)$

FIG. 4A

Coil #2

$W_2(1,1)$	$W_2(1,2)$	$W_2(1,3)$
$W_2(2,1)$	$W_2(2,2)$	$W_2(2,3)$
$W_2(3,1)$	$W_2(3,2)$	$W_2(3,3)$

FIG. 4B

Coil #3

$W_3(1,1)$	$W_3(1,2)$	$W_3(1,3)$
$W_3(2,1)$	$W_3(2,2)$	$W_3(2,3)$
$W_3(3,1)$	$W_3(3,2)$	$W_3(3,3)$

FIG. 4C
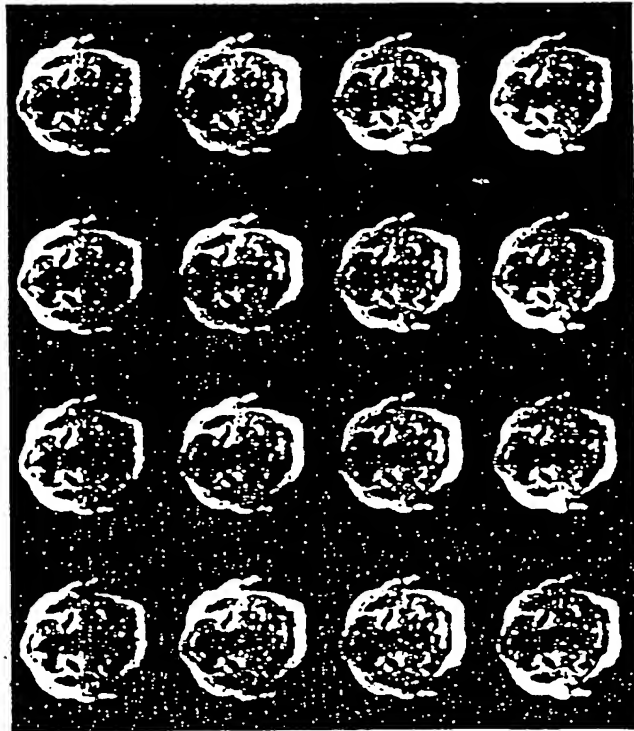
Figure 2: Simulation of the images received with a coil having the same sensitivity profile placed at 16 different positions around the brain .

Figure 2: Results of the reconstruction simulation using different sets of k-space lines. From left to right, the images were reconstructed by using the k-space lines acquired from skipping respectively 1,5,10 and 16 lines.
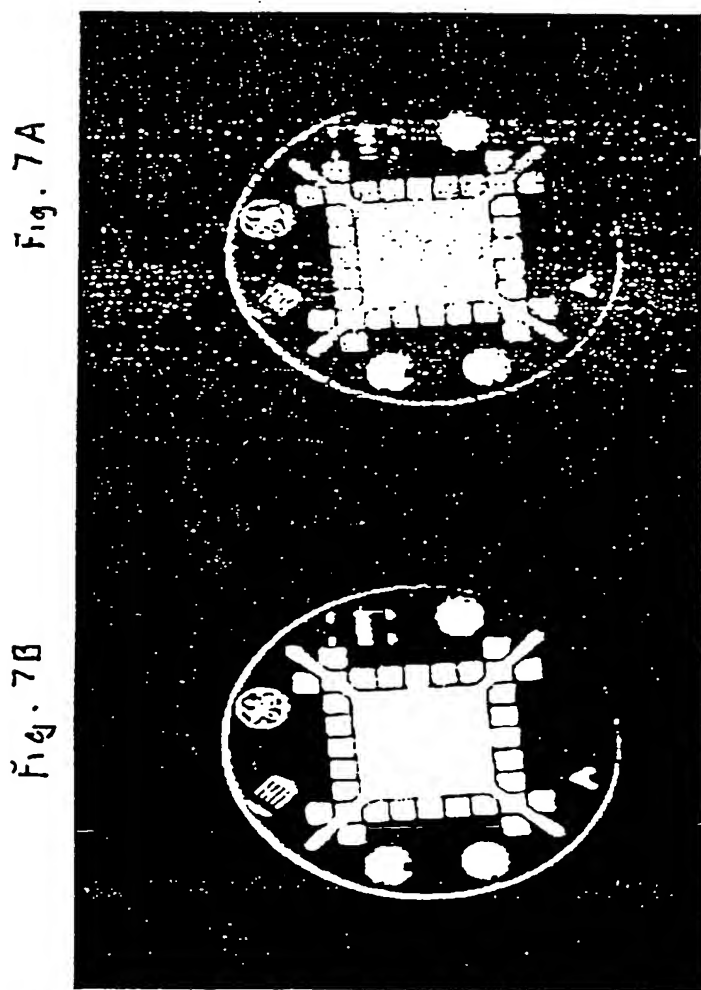
Fig. 7A

Fig. 7B

Figure 4. Experiment results: The left figure shows an image of a GE quality phantom aquired using a RARE sequence, and the image on the right shows the same image acquired with a 16 fold speed enhancement using our technique.

Fig. 8 A

Fig. 8 B
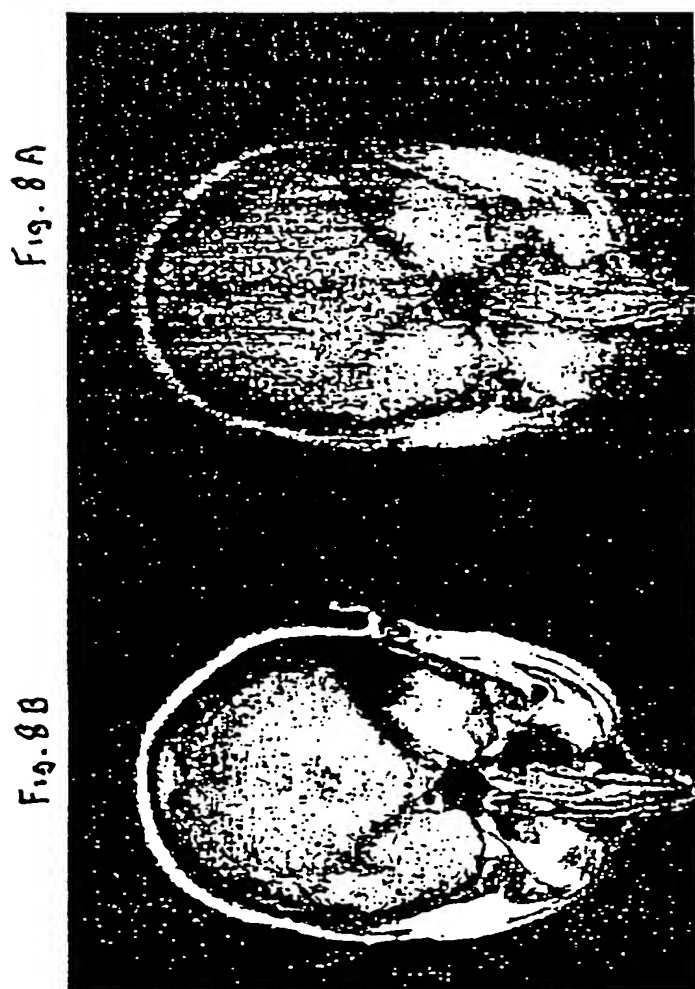
Figure 5: Experiment results: The left figure shows a brain image aquired using a RARE sequence, and the image on the right shows the same image acquired with a 16 fold speed enhancement using our technique.

INTERNATIONAL SEARCH REPORT

A.	CLASSIFICATION OF SUBJECT MATTER

IPC(7) :G01V 3/00
US CL   :324/307, 309

According to International Patent Classification (IPC) or to both national classification and IPC

B.	FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

U.S. : 324/307, 309, 314, 318, 322, 300

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

NONE

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

NONE

C.	DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	US 5,587,656 A (VAN HEELS-BERGEN et al.) 24 December 1996 (24.12.1996), fig. 2, col. 1, col. 2, col. 5 lines 30-65.	1-5
A	US 5,712,567 A (WANG) 27 January 1998 (27.01.1998), fig. 4 and abstract.	1-5
A,P	US 5,945,826 A (LEUSSLER) 31 August 1999 (31.08.1999), abstract.	1-5

☐ Further documents are listed in the continuation of Box C.   ☐ See patent family annex.

*	Special categories of cited documents:	"T"	later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"A"	document defining the general state of the art which is not considered to be of particular relevance		
"E"	earlier document published on or after the international filing date	"X"	document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
"L"	document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)	"Y"	document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art
"O"	document referring to an oral disclosure, use, exhibition or other means		
"P"	document published prior to the international filing date but later than the priority date claimed	"&"	document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
29 JUNE 2000	19 JUL 2000

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